ASSEMBLY AND PLANAR STRUCTURE FOR USE THEREIN WHICH IS EXPANDABLE INTO A 3-D STRUCTURE SUCH AS A STENT AND DEVICE FOR MAKING THE PLANAR STRUCTURE

CROSS-REFERENCE TO RELATED APPLICATION

This application claims the benefit of U.S. provisional application Serial No. 60/433,846, filed December 16, 2002 and entitled "Design and Fabrication of Stents Using Planar Metal Foils."

STATEMENT REGARDING FEDERALLY SPONSORED RESEARCH OR DEVELOPMENT

This invention was made with Government support under National Science Foundation Grant No. ECS-0233174. The Government has certain rights in the invention.

BACKGROUND OF THE INVENTION

1. Field of the Invention

This invention relates to assemblies and planar structures for use therein which are expandable into 3-D structures such as stents and devices for making the planar structures.

2. Background Art

The following references are noted hereinbelow:

[1] D.A. Leung et al., "Selection of Stents for Treating Iliac Arterial Occlusive Disease," J. VASC. INTERV. RADIOL., Vol. 14, 2003, pp. 137-52.

[2] P.E. Andersen et al., "Carotid Artery Stenting," J. INTERV. RADIOL., Vol. 13, No. 3, 1998, pp. 71-6.

[3] C.R. Rees, "Stents for Atherosclerotic Renovascular Disease," J. VASC. INTERV. RADIOL., Vol. 10, No. 6, 1999, pp. 689-705.

[4] C. Pena et al., "Metallic Stents in the Biliary Tree," MIN. INVAS. THER. & ALLIED TECHNOL., Vol. 8, No. 3, 1999, pp. 191-6.

[5] B.K. Auge et al., "Ureteral Stents and Their Use in Endourology," CURR. OPIN. UROL., Vol. 12, No. 3, 2002, pp. 217-22.

[6] Y.P. Kathuria, "Laser Microprocessing of Stent for Medical Therapy," PROC. IEEE MICROMECH. HUMAN SCI., 1998, pp. 111-14.

[7] K. Takahata et al., "Batch Mode Micro-Electro-Discharge Machining," IEEE J. MICROELECTROMECH. SYS., Vol. 11, No. 2, 2002, pp. 102-10.

[8] K. Takahata et al., "Coronary Artery Stents Microfabricated from Planar Metal Foil: Design, Fabrication, and Mechanical Testing," PROC. IEEE MEMS, 2003, pp. 462-5.

[9] J.C. Conti et al., "The Durability of Silicone Versus Latex Mock Arteries," PROC. ISA BIOMED. SCI. INSTRUM. SYMP., Vol. 37, 2001, pp. 305-12.

[10] R.C. Hibberler, "Mechanics of Materials Third Edition," PRENTICE-HALL, INC., 1997.

[11] S.N. David Chua et al., "Finite-Element Simulation of Stent Expansion," J. MATERIALS PROCESSING TECHNOL., Vol. 120, 2002, pp. 335-40.

[12] "Metals Handbook Ninth Edition," Vol. 8 Mechanical Testing, AMERICAN SOCIETY FOR METALS, 1985.

[13] F. Flueckiger et al., "Strength, Elasticity, and Plasticity of Expandable Metal Stents: In-Vitro Studies with Three Types of Stress," J. VASC. INTERV. RADIOL., Vol. 5, No. 5, 1994, pp. 745-50.

[14] R. Rieu et al., "Radial Force of Coronary Stents: A Comparative Analysis," CATHETER. CARDIOVASC. INTERV., Vol. 46, 1999, pp. 380-91.

[15] For example: J.D. Lubahn et al., "Plasticity and Creep of Metals," JOHN WILEY & SONS, 1961.

U.S. Patent Nos. 6,624,377 and 6,586,699 are related to the present application.

Stents are mechanical devices that are chronically implanted into arteries in order to physically expand and scaffold blood vessels that have been narrowed by plaque accumulation. Although they have found the greatest use in fighting coronary artery disease, stents are also used in blood vessels and ducts in other parts of the body. These include iliac arteries [1], carotid arteries [2], renal arteries [3], biliary ducts [4] and ureters [5]. The vast majority of coronary stents are made by laser machining of stainless steel tubes [6], creating mesh-like walls that allow the tube to be expanded radially with a balloon that is inflated during the medical procedure, known as balloon angioplasty. This fabrication approach offers limited throughput and prevents the use of substantial resources available for fabricating planar microstructures.

Micro-electro-discharge machining (μEDM) is another option for cutting metal microstructures. This technique is capable of performing 3-D micromachining in any electrical conductor with sub-micron tolerance and surface smoothness. It has not been extensively used for stent production in the past because traditional μEDM that uses single electrodes with single pulse timing circuits often suffers from even lower throughput than the laser machining. However, it has been recently demonstrated that the throughput of μEDM can be vastly increased by using spatial and temporal parallelism, *i.e.*, lithographically formed arrays of planar electrodes with simultaneous discharges generated at individual electrodes [7].

SUMMARY OF THE INVENTION

An object of the present invention is to provide an assembly and planar structure for use therein which is expandable into a 3-D structure such as a stent and device for making the planar structure wherein the planar structure can be readily manufactured.

In carrying out the above object and other objects of the present invention, a planar structure expandable into a 3-D structure is provided. The planar structure includes first and second spaced side beams which extend along a longitudinal axis. A plurality of spaced cross-bands connect the side beams together. A first set of the cross-bands are expandable in a first direction substantially perpendicular to the longitudinal axis to form a 3-D structure.

The side beams may be substantially straight and/or continuous.

A second set of the cross-bands may be expandable in a second direction substantially opposite the first direction to form a mesh-like 3-D structure.

Adjacent cross-bands may be expandable in the opposite directions to form a mesh-like 3-D structure.

The planar structure may plastically deform during expansion so that the 3-D structure is free-standing, or may have a cylindrical geometry.

The 3-D structure may be a tubular stent.

The planar structure may include a conductive foil.

Each of the cross-bands may include a series of folded beams.

The folded beams may have an involute pattern or a switchback pattern.

Each of the cross-bands may include hinges for interconnecting adjacent folded beams.

The side beams and cross-bands may include biocompatible surface coatings.

The side beams and cross-bands may be made of a biocompatible metal.

The cross-bands may be made of a shape-memory alloy, and the planar structure may be self-expandable.

The side beams and cross-bands may be made of at least one of a biocompatible and a biodegradable polymer.

The side beams and cross-bands may be formed by removing material from a sheet of material.

The sheet of material may include conductive foil, and the side beams and cross-bands may be formed by electric discharge machining the conductive foil.

At least the first side beam may include a link portion having a mechanical strength lower than other portions of the first side beam to allow the first side beam to break at the link portion during expansion of the first set of cross-bands.

The link portion may be thinned relative to the other portions of the first side beam.

The link portion may be made of a fragile material relative to the other portions of the first side beam.

The 3-D structure may be a helical coil.

The helical coil may include at least one electrical inductor.

The 3-D structure may comprise at least one electrical conductor.

The helical coil may include first and second spaced rings at opposite ends thereof. Each of the rings may be formed by an adjacent pair of expanded cross-bands.

At least the first ring may include a dielectric part which mechanically connects but electrically insulates adjacent portions of the first ring.

At least the first ring may include a link portion having a mechanical strength lower than other portions of the first ring to allow the first ring to break at the link portion during expansion of the first set of cross-bands to open an electrical path formed by the first ring.

At least one of the side beams and the cross-bands may include a dielectric part which mechanically connects but electrically insulates adjacent portions of the at least one of the side beams and the cross-bands.

Further in carrying out the above object and other objects of the present invention, an assembly including a planar structure is provided. The planar structure includes a pair of spaced side beams which extend along a longitudinal axis. First and second sets of spaced cross-bands connect the side beams together. A balloon is mounted on the cross-bands so that adjacent cross-bands are disposed on opposite first and second sides of the balloon. Inflation of the balloon causes the first set of cross-bands on the first side of the balloon to expand in a first direction and the second set of cross-bands on the second side of the balloon to expand in a second direction substantially opposite the first direction and substantially perpendicular to the longitudinal axis to form a mesh-like, 3-D structure.

The balloon may be an angioplasty balloon and the 3-D structure may be a tubular stent.

The assembly may further include a catheter tube in fluid communication with the angioplasty balloon.

Still further in carrying out the above object and other objects of the present invention, a device for use in a electric discharge machining system to form an expandable planar structure from a conductive planar workpiece is provided. The device includes a substrate and a planar electrode formed on the substrate. The planar electrode includes a pair of spaced, side electrode members extending along a longitudinal axis to form a pair of side beams of the structure from the workpiece. The planar electrode further includes a plurality of spaced cross-band electrode members to form a plurality of spaced cross-bands of the structure from the workpiece. The cross-bands connect the side beams together.

The side electrode members and the cross-band electrode members may include a plurality of copper structures formed by electroplating the substrate.

The substrate may include a semiconductor wafer. The side electrode members and the cross-band electrode members may include a plurality of semiconductor structures formed by removing material from the semiconductor wafer.

The above object and other objects, features, and advantages of the present invention are readily apparent from the following detailed description of the best mode for carrying out the invention when taken in connection with the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

FIGURES 1a-1c are generalized, schematic views of a planar structure or stent of the present invention, mounted on a deflated balloon and expanded to a cylindrical geometry, respectively;

FIGURE 2 is a detailed schematic view of a first embodiment (*i.e.*, design) of a planar structure of Figure 3 mounted on a deflated angioplasty balloon in fluid communication with a catheter tube;

FIGURE 3 is a top plan schematic view of the first embodiment of the planar structure having a pattern of folded beams in an involute shape;

FIGURE 4 is a top plan schematic view of a second embodiment of the planar structure having a pattern of folded beams in a switchback shape;

FIGURE 5 is a schematic view of an electric discharge machining system including an electrode for cutting the planar structure of Figure 1a;

FIGURE 6 is a perspective schematic view of the system of Figure 5 with a device including a planar electrode formed on a substrate for cutting a workpiece to form the planar structure of Figure 1a;

FIGURE 7a is a top plan schematic view of an embodiment of the planar structure;

FIGURE 7b is a close-up view of a thinned link of a side beam of the planar structure of Figure 7a;

FIGURE 8a is a top plan schematic view of another embodiment of the planar structure;

FIGURE 8b is a close-up view of a fragile plug embedded in a side beam of the planar structure of Figure 8a;

FIGURES 9a-9d are perspective, schematic views showing a planar structure having breakable links by itself, on a deflated balloon, on a fully expanded balloon, and its electrical path equivalent to the final stage of Figure 9c,

respectively, wherein the resulting 3-D structure includes a helical coil with end rings;

FIGURE 10a is a top plan schematic view of another planar structure having breakable links;

FIGURE 10b is a simplified view of a 3-D structure in the form of a pair of helical inductors formed after the expansion of the planar structure of Figure 10a;

FIGURE 11a is a perspective, simplified view of an expanded 3-D structure having end rings;

FIGURE 11b is a close-up view of plugs of a dielectric material embedded in the end rings of Figure 11a;

FIGURE 12a is a perspective, simplified view of an expanded 3-D structure having end rings;

FIGURE 12b is a close-up view of breakable links or portions of the end rings of Figure 12a;

FIGURE 13a is a top plan schematic view of another embodiment of a planar structure;

FIGURE 13b is a close-up view of an interconnecting dielectric plug used in the planar structure of Figure 13a; and

FIGURE 14 is a perspective, schematic view of folded beams having an involute shape in initial (indicated by dashed lines) and expanded (indicated by solid lines) positions.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

A new assembly and planar structure for use therein which is expandable into a 3-D structure such as a stent and device for making the planar structure are disclosed herein. Also, this invention presents a new approach to the design and manufacture of coronary artery stents, which permits the use of planar batch fabrication techniques using microelectrodischarge machining. The devices are compatible with standard stenting tools and procedures. The wall patterns were designed so that both stress relief and the mechanical strength are simultaneously achieved in the expansion.

Referring to Figures 1a-1c, a generalized schematic view of a stent or planar structure constructed in accordance with the present invention is generally indicated at 10. The stent 10 is mounted on a deflated balloon 12 in Figure 1b and on the balloon 12 when inflated in Figure 1c. The stent 10 includes involute bands 16 tied between a pair of side beams 14. Measurements demonstrate that the designs have the same radial strength as a commercial stent even though the former use metal that is only about half as thick. The thinner walls also contributed to achieving higher longitudinal flexibility than a commercial one in the expanded state. Both the radial strength and the flexibility are found to have no significant dependence on orientation relative to the original planar direction of the foil. Dimensional variations in tubular diameter, longitudinal shrinkage, and radial recoiling in the expanded stents are at most a few percent.

The invention will also facilitate other three-dimensional structures such as antennas and transformers. Using this approach, any electrically conductive material can be used to form a tubular mesh-like structure. This includes structures which have attached elements that do not conform to the shape of the cross-section of the tube, such as tangential cantilever or loop attached to the perimeter. The structures can be used as inductors (*i.e.*, Figure 10b), antennas, transformers, or capacitors for electrical circuits. They may also be used for mechanical functions such as springs, trusses, etc. in microsystems.

The new fabrication approach uses metal foils as starting materials for the stents 10, which permits the parallelism described in U.S. Patent Nos. 6,624,377 and 6,586,699 to be exploited, thereby offering high throughput and repeatability. The favored mechanical characteristics including radial strength and longitudinal compliance in expanded stents (*i.e.*, Figure 1c) have been experimentally and theoretically investigated, and are discussed with comparisons to commercial stents.

A variation that uses strategically located breakable links (Figures 7a, 8a, 9a, 10a) in the stent provides additional freedom in customizing the mechanical and electrical properties of these devices.

Design and Fabrication

The fabrication approach was applied to μEDM 50 μm-thick stainless steel foil into a planar structure, generally indicated at 20, that could be slipped over an angioplasty balloon 22 and be reshaped into a cylinder when deployed in the manner of a conventional stent via a catheter tube 28, as shown in Figure 2. The planar pattern of the structure 20 provides the important mechanical characteristics of radial stiffness and longitudinal compliance in the expanded structure. In order to reduce the likelihood of joint failure, it was decided to develop a structure 20 that was completely flexural in nature, and did not have any bonded or hinged joints. This effort used 50 μm-thick type 304 steel which is very similar to the 316 steel commonly used for commercially available stents.

Several layouts were designed and experimentally tested. The best results in terms of mechanical characteristics (discussed herein below) were obtained with the design shown in Figures 2 and 3, which is referred to as design 1 (*i.e.*, units are in μm). The pattern has two longitudinal side beams 24, which are connected transversely by cross-bands 26, each of which contain three identical involute loops (*i.e.*, Figure 14 shows one such loop). The involute shape is tailored to provide selected stress-relief during expansion of the stent 20 to the desired deployment diameter, which is 2.65 mm in this case. In order to increase radial

strength, this design uses a larger number of cross-bands 26 per unit length of the stent 20 and beams A_n, C_n, and E_n are designed to be longer than the others, B_n and D_n.

Another representative planar substrate is illustrated at 40 in Figure 4 (*i.e.*, units again in μm), which is referred to as design 2. It has similar dimensions and a configuration that uses an array of cross-bands 42 and two side beams 44, but the cross-bands 42 have a switchback pattern in this case. In contrast to the involute design, the beams in segment G, which are parallel to the longitudinal axis, are longer than the others in segment H, which are perpendicular to the axis. This design, in general, has a higher expansion ratio to the initial width between the side beams 44 in radial direction, but fewer cross-bands 42 along the longitudinal axis.

To emulate the deployment of a stent, the angioplasty balloon 22 was threaded through the 7 mm-long planar structure 20, as shown in Figure 2, such that the transverse bands 26 alternated above and below it. With the set-up illustrated in Figures 1b and 2, the stent 20 was expanded by inflating the balloon 22 with liquid up to 12 atm. pressure, in a manner identical to commercial stents, resulting in the structure similar to the one shown in Figure 1c. Variation in the diameter of expanded stents was typically within $\pm 4\%$, while radial recoil upon deflation of the balloon 22 was even smaller than that. The shrinkage in length upon the expansion was $<3\%$. A deployment inside a mock artery was done. The mock artery used was a commercially available silicone-based tube (Dynatek Dalta Scientific Instruments, MO, USA) with 3 mm diameter and 0.25 mm wall thickness, which is tailored to have radial compliance comparable to human coronary arteries [9]. In this deployment, the stent 20 was expanded to 3.5 mm diameter. The tube had a distended sidewall at the location where the stent 20 was deployed, demonstrating mechanical strength large enough to prevent the relaxation of the simulated artery.

Upon expansion of the stent, beams in the structure are permanently deformed as shown in Figure 14. The pattern of the stent must, therefore, be designed to accommodate large deformations so that the maximum tensile stress is

less than the ultimate stress, which is about 517 MPA for the 304 stainless steel [10]. The deformation and resultant stresses were evaluated by using an FEA package, ANSYS™. The simulation used a bilinear stress-strain model, and the following mechanical properties of the steel [10,11]: Young's modulus = 193 GPA, yield stress = 207 MPA, tangent modulus = 692 MPA, and Poisson's ratio = 0.27. Figure 14 shows a unit involute section, generally indicated at 140, of the cross-bands of design 1 with a displacement that approximately corresponds to the deployed diameter. The section 140 includes beams 142 with interconnecting hinges 144. The maximum von Mises stress appears at the location indicated near the flexural hinge element B_n and is 382 MPA, sufficiently below the ultimate stress.

In addition to the bending of beam segments, torsional deformations also play important roles in expanding a stent and maintaining its final shape. The most significant ones are in the side beams 24, which are twisted by 90-180° along the segment *F* (labeled in Figure 3) between two adjacent bands 26. Different torsional deformation was observed at a flexural hinge *H* in design 2 (*i.e.*, Figure 4). The approximate shear strain for both these cases was shown on a shear stress versus shear strain response curve for 304L stainless steel obtained from [12]. It was evident that beam fracture associated with only the torsion is not a concern for the stent. For the test, the hinge as well as the beams had 50 μm square cross-section. Although the strain due to this torsion is well below the fracture point, additional deformations at the site also include bending that may further increase the maximum strain experienced. Mechanical failure was observed due to a combination of severe bending and tension. This fracture was observed in design 1_A, a precursor to design 1 for which width of flexural hinges was 50 μm, and segments A_n *(and* E_n*)*, B_n *(and* D_n*)*, and C_n were 550, 150, and 450 μm, respectively. The narrower width and shorter length in the flexural hinges, B_n and D_n, of this design contributed increasing the tensile stress at the hinge. Since this was the only failure experienced, it is likely that an instance of metallurgical defect may have contributed to it. In design 1, a larger safety margin was incorporated by two changes: (i) doubling the widths of the segments B_n and D_n from 50 μm to 100 μm, and (ii) increasing the lengths of the same segments from 150 μm to 200 μm by doubling the gap between adjacent beams, as seen in Figure 3.

In addition to the stent fabrication, the planar scheme can be easily extended to fabrication of 3-D inductors, generally indicated at 108 in Figure 10b. The final 3-D structure is essentially a set of series-connected rings which offer negligible inductance. Use of breakable links in planar structures 70, 80, 90 and 100 (Figures 7a, 8a, 9a, and 10a) permits formation of helical coils or inductors 98 and 108 (Figures 9d with electrical connection 99 and 10b, respectively) in the same manner for the deployment of stents. Figures 7a and 7b show thinned links 78. Figure 9a shows thinned breakable links 94 in the planar structure 90 including cross-bands 96. Figures 8a and 8b show fragile plugs 88 to make its links breakable.

When a balloon 92 is inflated for expanding the planar structure 90 (Figure 9c), torsional strain developed in the side beams 91 is effectively concentrated at the links 94 made in the beams 91 (Figure 9a), leading to fracture (Figure 9c). The resultant final shape can be helical by placing the links 94 at selected locations. This fracture is controlled breakage, and the fractured cross-section area is minimal.

In like fashion, torsional strain developed in side beams 101 (*i.e.*, Figure 10a) is effectively concentrated in breakable links 104 in the beams 101 leading to fracture as shown in Figure 10b with electrical connection 109.

Experimental Results

The radial strength is a paramount mechanical characteristic in the stents. Several past efforts have assessed the strength in commercial stents [13,14]. To evaluate the devices of the present invention, short samples for involute and switchback designs were prepared and subjected to loading tests in which the reaction force per unit length of the stent is measured as a function of radial deformation . A sample is held in a groove mounted on the stage and compressed toward the probe. The gauge is rigidly fixed, and the displacement of the gauge probe is negligible compared to that of the sample. The force was measured by a gauge (Imada, Inc., IL, USA, DPS-1) that provides 1 mN resolution while first

compressing the stent by 1.5 mm in 25 μm increments, and then while relaxing the deformation.

A commercial stent with 316 stainless steel of thickness varying over 90-130 μm was tested for comparison. Measurements demonstrate that the design that uses the involute cross-bands (design 1_A) has the same radial strength to the commercial stent with similar diameter and twice the thickness. In addition, it exhibits better elastic recovery after loading, which suggested that it has better radial elasticity but the same stiffness as the commercial one. The switchback pattern (design 2), which as fewer cross-bands per unit longitudinal length, provides less radial strength than the involute pattern.

Orientation dependence of the radial strength was a concern since they were shaped from planar sheets as shown in Figure 6. Identical samples with four cross-bands of design 2 were tested at two different orientations: (A) perpendicular to a plane that includes both side beams, and (B) parallel to the plane. The measurements demonstrate that the radial strength is similar in both cases.

The experimental results showed a few discontinuities in the response curve. As can be seen in Figure 14, beams that correspond to $\dot{C}_n$ in Figure 3 are designed to rotate about their center by ~90° during the expansion. As a result, hinges D_n and B_{n+2} are positioned closely to each other. In addition, alternate cross-bands 26 in Figure 3, which adjoin each other when they are mounted on the balloon 22, deform in a way that the gaps between their segments are reduced as the stent 20 expands since the side beams 24 are deformed to wave-like shapes. The combination of these effects results in increased probability of physical contact between the hinges D_n and B_{n+2} as the balloon 22 is being inflated. As loading is applied, hinges happen to come into contact and get intermeshed, and then snap apart as the loading is further increased. This particular sample, being design 1_A, had a reduced gap of 50 μm between the cross-bands 26, which could also contribute to increase the probability. This undesirable mechanical interaction however can be improved by optimizing the layout.

Longitudinal compliance is a favored characteristic in stents. This is because the stent, fitted on an angioplasty balloon in a state that is only slightly expanded, must often travel a convoluted path along a blood vessel in order to reach the location of the deployment. In addition, longitudinal flexibility in a fully expanded stent can be beneficial for its deployment at curved sites. The longitudinal compliance of the fabricated stents was tested. A fully-expanded 7 mm long stent of design 1 was attached to a holder such that a 4 mm segment out of it was overhanging and unsupported. Using a force gauge, the displacement response was plotted for an end load. A similar test was also applied to the commercial stent tested before. The results reveal that the stent of the present invention had spring constants of 50 N/m and <5 N/m depending on the orientation, whereas that in the commercial stent resulted in 515 N/m. While this test was only performed on expanded stents, it suggests that the stents of the present invention perform favorably in this respect.

The design and fabrication of coronary artery stents of the present invention is preferably based on use of planar stainless steel foil and μEDM technology, as generally shown in U.S. Patent No. 6,624,377. An electrode 52 is controlled by a control unit in Figure 5. Figure 6 shows a workpiece 64 processed by a device including a substrate 66 on which is formed a planar electrode, generally indicated at 68, having side and cross-band electrode members 60 and 62, respectively.

The devices are intended to be compatible with standard stenting tools and procedures. The wall patterns were designed using FEA so that both the stress relief and the mechanical strength are simultaneously achieved in the expansion. The devices include involute bands tied between a pair of side beams. Measurements demonstrate that the designs have the same radial strength as a commercial stent even though the former use metal that is only about half as thick. The thinner walls also contributed to achieving at least 10X higher longitudinal flexibility than a commercial one in the expanded state. Both the radial strength and the flexibility are found to have no significant dependence on orientation relative to the original planar direction of the foil. Dimensional variations in tubular diameter,

longitudinal shrinkage, and radial recoiling in the expanded stents are at most a few percent.

All devices tested in this effort were fabricated by batch-compatible μEDM, which can open a path to exploit photolithography-based fabrication resources for the stent production [7]. As an extension of this technology for manufacturing stents, use of strategically-located breakable links as described above also facilitate fabrication of other 3-D structures such as antennas and transformers.

Furthermore, referring to Figures 13a and 13b, dielectric plugs 132 may be incorporated into a planar structure 130 to ensure that electric current does not attempt to flow in end rings in the expanded 3-D structure (not shown).

Figures 11a and 11b show an expanded 3-D structure in the form of a helical coil 110 having end rings 112 with such dielectric embedded plugs 114.

Figures 12a and 12b also show an expanded 3-D structure in the form of a helical coil 120 having end rings 122 with thinned portions 124 so that the end rings 122 are breakable to prevent current flow in the end rings 122.

While embodiments of the invention have been illustrated and described, it is not intended that these embodiments illustrate and describe all possible forms of the invention. Rather, the words used in the specification are words of description rather than limitation, and it is understood that various changes may be made without departing from the spirit and scope of the invention.